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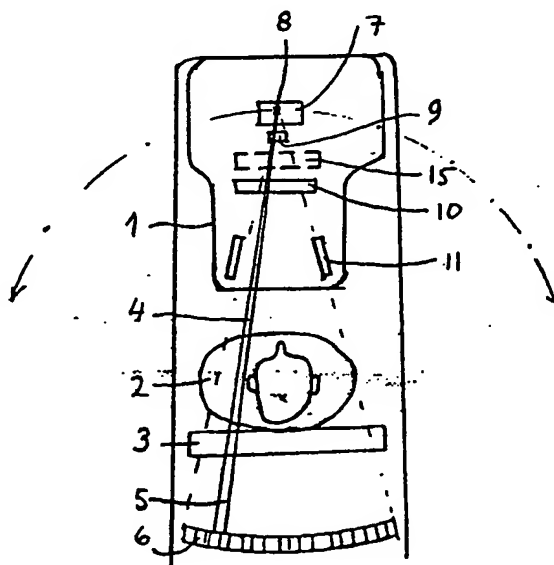
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54 **CT scanner for radiation therapy planning.**

57 A device for making photon CT images with the patient in radiation therapy position. A narrow pencil shaped beam of photons (5) the energy of which is of the order of about 3 up to about 50 MeV is scanned transversally over the patient. During scanning the photons transmitted through the patient's body are measured by the detector array (6). A computer (13) then reconstructs a CT-image of the body from the photon transmission data and the position data of the scanning beam (5). The CT image thus determined is used for planning the patient treatment in a separate dose planning computer (16). The patient is then treated by scanning the beam over the treatment volume and in so doing using the recorded position data to position the beam and the absorption data to vary the scan pattern of the beam such that the desired dose distribution in the tumor volume is obtained.



A CT SCANNER FOR RADIATION THERAPY

5 The present invention relates to a method and device of making
photon CT images with the patient in the radiation therapy
position by subjecting the patient to the therapy radiation
beam. The photon radiation used for taking the CT images is
also used for the therapy radiation. At first the CT images
are taken and are computer processed and then, some minutes
later while the patient still is resting on the patient couch,
the radiation therapy is performed. This method is generally
10 known from Med.Phys.9(4), July/Aug 1982, Am. Assoc. Phys. Med.
wherein R.G. Simpson et al. in a technical report entitled "A
4-MV CT scanner for radiation therapy: The prototype system."
is describing the novel concept referred to above.

15 In the known system a flat fan like photon beam is directed
towards the patient and a photon detector array, placed under
the patient's couch, is providing electrical signals
representing the photon transmission along a sectional view of
the patient's body. The gantry, which provides the fan like
photon beam, is rotated in small steps around the patient's
20 body and for each gantry setting the electrical signals are
stored in a computer which out of the data so collected forms
a CT image of the sectional view of the patient. The main
advantage with this known concept is that no separate
diagnostic-energy CT scanner is required. Moreover, the
25 contrast to the images obtained with the known 4-MV CT scanner
is improved, particularly the contrast to body tissue, Also
the dose planning is more accurate. Previously the attenuation
coefficients of the tissue, taken with diagnostic radiation
energy of typically 70 keV has to be extrapolated to those
30 expected to be present at the therapy radiation energies,
which typically may be in the order of about 5 MeV. An
extrapolation over such a wide range is uncertain.

Before the Simpson concept was desired conventionally the
therapy treatment was preceded by an X-ray screening -
35 performed by a conventional CT-scanner - in order to localize

the tumor volume. Thereafter the patient was moved to the couch of a therapy apparatus the radiation field of which was set (by a collimator) on basis of the CT-image. Apparently the coordinate system of the therapy apparatus is different from what of the CT-scanner making it difficult to exactly position the therapy apparatus on the tumor volume. With the Simpson concept such positioning arrows of the tumor volume are obviated.

The Simpson CT-scanner referred to above have some inherent drawbacks among which the following are apparent:

- The flat, broad fan like beam radiates a "slice" of the patient's body simultaneously implying that a detector included in the detector ray under the patient's couch further to electrones which pass the volume immediately above the detector and secondary electrones generated by said volume also will detect secondary electrones generated in the volume surrounding said volume immediately above the said detector. This degrades the resolution of the final CT-image.

- Dynamic radiation treatment of the tumor volume to the desired dose rate is difficult to perform. While rotating the gantry of the treatment apparatus the field collimator must be continuously readjusted to adapt its field to the shape of the tumor volume as seen from the gantry.

- The radiation is of a specific energy (4 MeV). Should a different energy be used then a different beam flattening filter must be used. Such beam flattening filters further to being large and heavy inevitable absorbs the high energy electrones included in the primary beam from the accelerator.

The present invention is directed to a method and apparatus for reducing the deficiencies inherent in the known Simpson CT-scanner, particularly:

- The resolution of the CT-image should be improved, - dynamic

radiation treatment should be possible to perform in an easy way, - a spectrum of different radiation energies should be available thereby making it possible to take CT-images at a desired beam energy without using beam flattening filters which inevitable absorbs the high energy electrones, that is exactly those electrones which in accordance with the present invention should be used to make the CT images.

The above objects of the invention are achieved with a method and device as defined in the accompanying patent claims.

A preferred embodiment of the invention should be described in conjunction with the accompanying drawings wherein:

Fig. 1 is a diagram showing the absorption cross section for different tissues versus the energy of the photon beam,

Fig. 2 is an end view of the apparatus in accordance with the present invention,

Fig. 3 schematically shows the electrical signal obtained at the photon detector array for the specific position (Fig. 3a) of the scanned beam in accordance with the present invention, and for the next following position (Fig. 3b) of the scanning beam,

Fig. 4 is a corresponding electrical signal obtained at the photon detector of the known Simpson device and this figure has been incorporated in order to illustrate the differences between the known device and that of the present invention,

Fig. 5 illustrates the format in which position and photon transmission data are stored in the memory of a micro computer, and

Fig. 6 illustrates a block diagram of the apparatus in accordance with the present invention.

In Fig. 1 the absorption curves for bone, muscles and fatty tissues respectively are shown for varying photon energies. When the difference between the attenuation of the radiation in e.g. bone and fatty tissues is large, indicated with vertical double edged arrows, then the contrast of the CT image taken at the indicated radiation energy, is large. The larger the contrast the better the image quality. From Fig. 1 and the left vertical arrow it is apparent why conventional CT scanners are operating with an acceleration voltage of less than about 1 MV. From the right vertical arrow in Fig. 1 it is also apparent that for radiation energies over about 50 MV the contrast is high. The Simpson concept referred to above did take advantage of this fact. Moreover, the dose planning, as suggested by Simpson, was more accurate since the CT image taken with this high energy radiation showed exactly how the organs within the body absorbed the radiation and from this CT image it was possible to compute the real attenuation coefficients of the organs without relying on extrapolation from 70 kV CT images to therapy radiation energies in the order of 4 MV and above. Such extrapolation is commonly done with the aid of curves similar to that shown in Fig. 1.

In Fig. 2 there is shown a principal set up of the apparatus in accordance with the present invention. From a gantry 1 a patient 2 resting on a patient's couch 3 is radiated with a pencil like, well united, beam 4 of photons, some of which 5, penetrates the body and are detected by a detector array 6.

A proton beam generated by an accelerator not shown, e.g. a race track microtron, enters the gantry 1 which comprises a pair of scanning electromagnets 7 which are energized by a varying dc-current in order to scan the proton beam in the plane of the drawing between the end of positions shown in broken lines. The scanning system is preferably of the kind described in our U.S. Patent Serial No. 4 442 352. Generally at the scanning centre 8 of the proton beam there is a target 9 of tantalum or other material known per se which produces photons of generally the same energy as that of the proton

beam. Further down in the gantry there is a transmission ion chamber 10 measuring the beam current and a pair of collimator blocks 11 for beam collimation.

5 The beam is pulsed which means it comprises pulses of about 1-5 μ duration with a repetition frequency of 50-250 MHz. Due to the pulse nature of the beam the detector array is of integrating type, that is rather than counting each photon requiring a very fast responding detector it integrates the number of photons which are detected over a time period. The
10 detector array comprises a linear array of a number of Bismuthgermanate crystals arranged in a side by side relationship and each optically coupled to a photodiode which delivers an electrical signal proportional to the said integral, which in turn therefore is proportional to the
15 transmission of the radiated body tissue.

In Figure 3A there is shown the detector array 6 and its separate detectors x_i . In the shown embodiment the physical length of the detector is about 60 cm comprising 128 detectors each made of a crystal Bismuthgermanate allowing for a picture resolution about 5 mm which is a beam which is about 2 cm
20 in square. When the beam is at crystal x_n at time t_n the photon transmission has a value of T_n . In Figure 3B the beam has moved to x_{n+1} at time t_{n+1} and the transmission is T_{n+1} . The transmission values are arranged within a grey scale comprise 16 grey levels 0 corresponding to white and 16 to
25 black. From the above it is clear that the linear array 6 is used as a coordinate system for the beam. For each position x_i there is a transmission value T_i which is allocated x_i . The position and transmission data for each scan of the beam over the array is recorded in a micro computer which preferably is
30 of the 16 bit type; allowing convenient format for storing said data. This format is shown in Figure 5 from which it appears that high order byte stores the position of the beam while the low order byte stores the corresponding transmission values. After each scan of the beam over the array the gantry
35 is rotated together with the array through a predetermined

angel around the patient's couch which is at rest and a new scan is made and new position and transmission data values are recorded. This is repeated for each angular setting of the gantry until finally a completed 360° rotation of the gantry has been made. The data so collected and recorded is processed in a manner known per se in a frame computer shown at 12 in Figure 6 in order to reconstruct a CT image of the "slice" of the patient's body just irradiated. Thereafter either the patient's couch or the beam or both are moved a predetermined distance in the longitudinal direction of the couch and a new CT image of the body is taken while rotating the gantry around the patient's body.

Instead of assigning a transmission value T_i to each of the positions x_i by taking a separate reading at each detector which for example is done by using a multiplexer the output of which is connected to a scan and detector computer 13 shown in Figure 6 connected to a common output conductor via an A/D converter 14, the outputs of each of the detectors in the array one connected to a common conductor connected to said scan and detector computer 13; said common conductor in this case carrying a composite signal of the kind indicated in Figure 4.

For reference and for monitoring the incident photon beam a second photon detector 15 is inserted between the target 9 and the transmission ion chamber 10. Should the photon fluence vary during the scanning, the transmission values are correspondingly modified.

In accordance with the present invention the energy of the photons should be in the order of 50 Mev in order to produce pair production which gives a significant improvement in the contrast of the CT image.

From the quotient of the incident photon fluence to the recorded transmission value the attenuation coefficient is determined for each position x_i of the beam. This is made in a

separate dose planning computer 16. The tumor is identified either by inspection of the CT-images taken by the apparatus in accordance with the present invention or by inspection of CT-images taken by a conventional (70-140 kV) CT-scanner. In both cases the dose is then planned with the aid of the dose planning computer 16.

If the dose is planned with the aid of a conventional CT-scanner image then the latter image is copied within the frame computer on the photon CT-image taken with the apparatus in accordance with the present invention. The resultant image will thus contain - for each pixel - (picture element of the image)

1) The position of each pixel (and therefore the position of the corresponding part of the patient's organ/tissue) expressed in the coordinate system of the CT scanner in accordance with the present invention and

2) the absorption coefficient of the corresponding part of the patient's organ(tissue, that is how much of the irradiation beam that is absorbed in different parts of the patient's body.

This will improve the dose planning since no extrapolation of the kind referred to above is needed. Moreover, no positioning errors will encounter. The dose planning is a matter of minutes and during this time the patient remains resting at the couch.

The radiation may also very easily perform dynamic radiation treatment of the target volume. In Figure 7 the target volume, for example a part of the throat of a patient, is shown at 17. The target volume should be treated with a constant dose rate. The overlying tissue, shown at 18, has varying thickness (shown at D1 och D2 respectively).

The incident radiation, shown at the vertical arrows, will therefore be absorbed in different rates in the tissue 13 leading to an uneven dose rate in the target volume 17. The

broken line 19 indicates the position of the dose maximum should the target 17 be radiated with a constant dose. As appears the maximum dose is situated beneath the target volume 17 when the position D2 is radiated from above. This is unwanted, since you want the dose maximum to lie at the same level as in portion D1. To compensate for this a smaller dose should be given in the thinner tissue D2 than in the thicker D1. By varying the speed of the scanning beam so that it is residing a longer time in the thicker tissue the dose rate in the target volume 17 is made constant over the target volume. For how long the beam should reside for example in the tissue is determined from the computed absorption coefficient for the corresponding portion of the thicker tissue and is made under program control from the dose planning computer.

CLAIMS

1. A method of making photon CT images with the patient in radiation therapy position by subjecting the patient to the therapy position by subjecting the patient to the therapy radiation beam, comprising the steps of:

- scanning a narrow pencil shaped beam of photons or neutrons the energy of which is of the order of about 3 up to about 50 MeV transversely over the patient,
- during said scanning measuring and recording - with a photon detector array located under the patient - the photons transmitted through the patient's body,
- allocating the current position of the scanning beam to the photon transmission - as measured - at said position,
- processing the data so recorded in order to form a photon CT image comprising the photon absorption properties of the patient's body at the photon energy used,
- using the photon absorption data CT image thus determined for planning the patient treatment,
- irradiating the patient by scanning the beam over the treatment volume and in so doing using the recorded position data to position the beam and said absorption data to vary the scan pattern of the beam such that the desired dose distribution in the tumor volume is obtained.

2. A method in accordance with claim 1, characterized by copying the absorption data CT image on an image formed by conventional CT apparatus and during said copy using the coordinate system of the recorded position data of the absorption data CT image to obtain accurate absorption data of each point of the tumor to be irradiated, and using the information so obtained during the scanning of the treatment beam to allocate each point of the tumor volume a desired irradiation dose.

3. A method in accordance with claim 2, characterized in that said scan pattern is varied by varying the speed of which the beam is scanned over the tumor volume during the radiation

teraphy.

4. A device for making photon CT image with the patient in the radiation therapy position by subjecting the patient to the therapy radiation, comprising:

5 - an accelerator producing a beam of high energy electrons,
- an electromagnetically operated beam scanning system for scanning the beam at least in a transverse direction over the patient's body,

10 - a target for the electron beam, said target being arranged generally at the scanning centre of the scanning system so as to produce a narrow, pencil formed beam of photons,

- a photon detector array positioned under the patient's body,
- first signal processor means connected to the photon detector array for producing first electrical signals representing the photons transmitted through the target volume at each current position of the scanned photon beam,

15 - second signal processor means for producing second electrical signals representing the current position of the scanning beam,

20 - computer means to store data relating to the photon attenuation of the target volume and obtained from said first electrical signals, to store data relating to the position said beam had when each first signal was generated,

25 - and frame computer means for planning the dose at each position of the target volume using said attenuation and position data and for controlling the movement and the speed of the scanned beam during therapy radiation by using said dose data and said previously stored position data.

30 5. A device in accordance with claim 4, characterized in that the energy of the photons is in the order of about 3MeV up to about 50 MeV.

35 6. A device in accordance with claim 4, characterized in that the photon detector array comprises bitmuthgermanate crystals arranged in a side by side relationship along a curved surface the centre of curvature of which coincides with

the scanning center of the scanning system.

7. A device in accordance with claim 6, characterized by means for moving the detector array in the longitudinal direction of the patient.

8. A device in accordance with claim 4, characterized by the fact that the second signal processor means comprise current sensors connected to an ion transmission chamber inserted between the target and a collimator included in the gantry which also houses the scanning magnets of the scanning system for the beam.

9. A device in accordance with claim 4, characterized in that the energy of the photons is high enough to produce pair generation effect in the tissue of the tumor volume.

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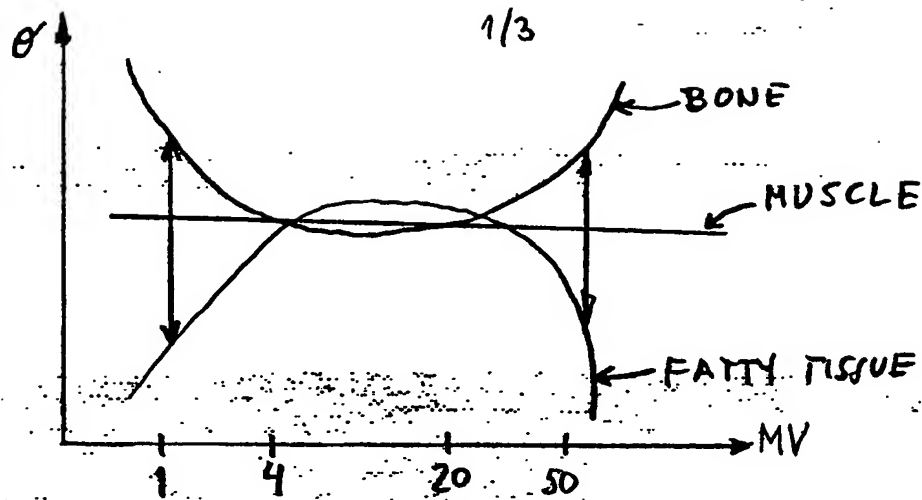


FIG 1

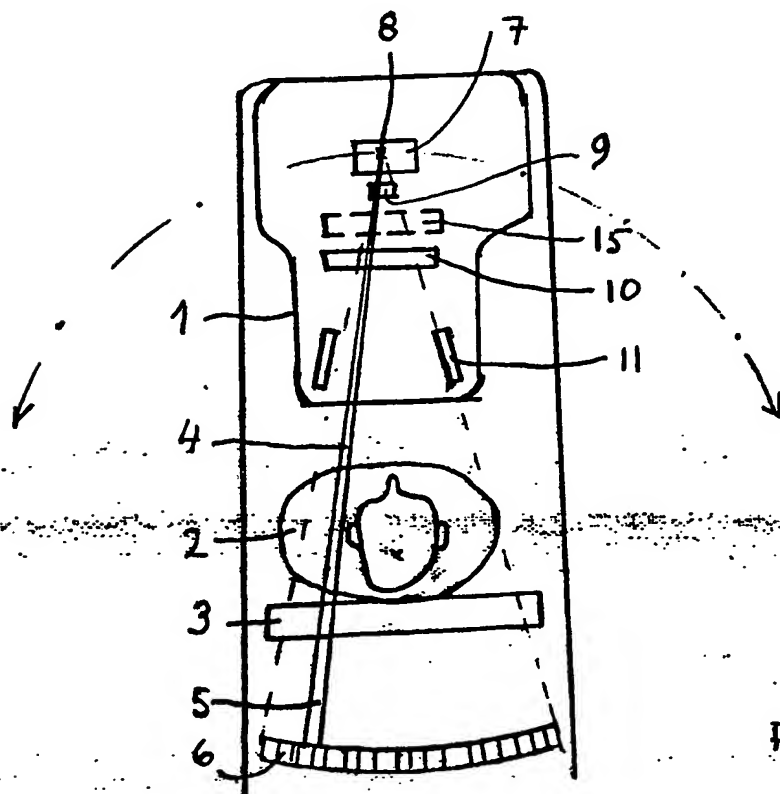
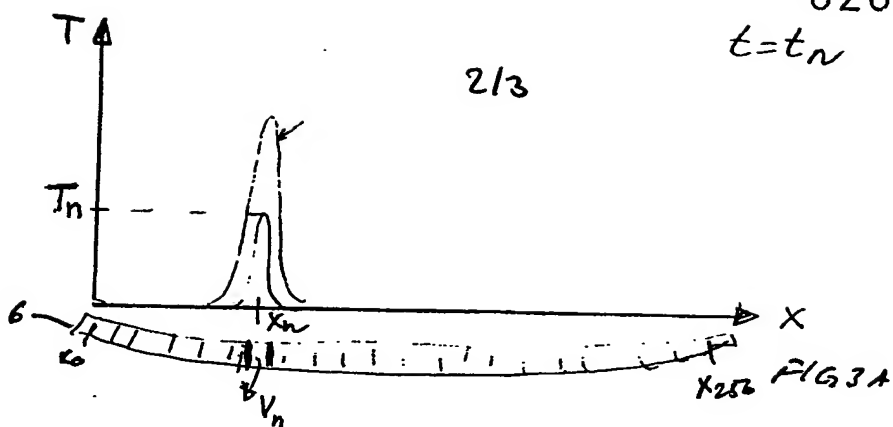


FIG2

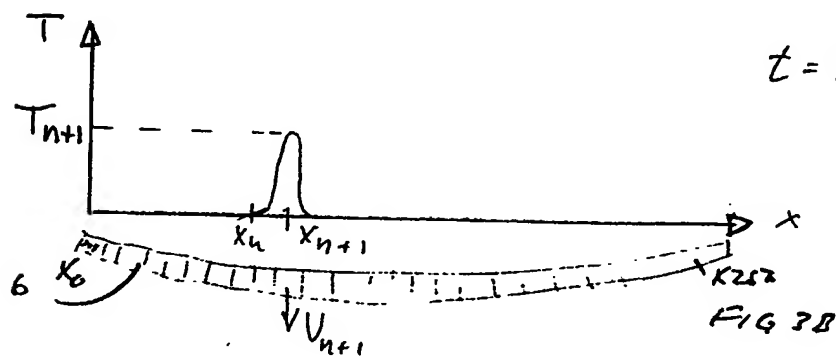
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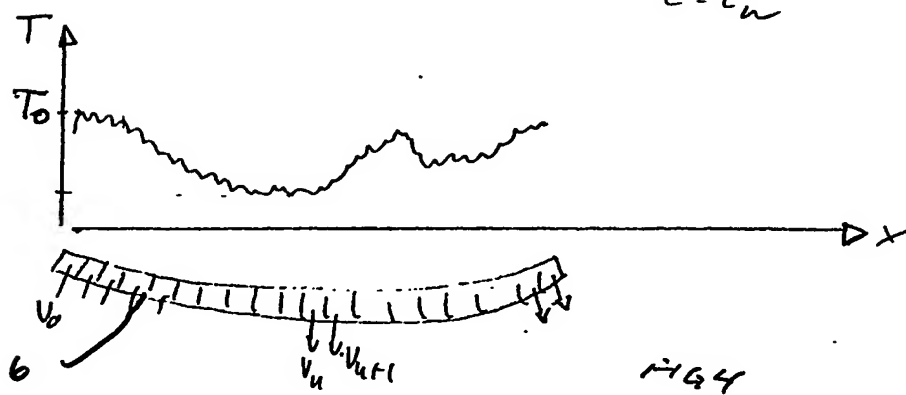
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PARTIAL EUROPEAN SEARCH REPORT
which under Rule 45 of the European Patent Convention
shall be considered, for the purposes of subsequent
proceedings, as the European search report

Application number

EP 85 85 0224

DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
D, Y	MEDICAL PHYSICS, vol.9, no.4, July/August 1982, pages 574-579; New York, US R.G. SIMPSON et al.: "A 4-MV CT scanner for radiation therapy: The prototype system."		A 61 B 6/03 A 61 N 5/10
A	* Whole article *	1,4 3	
Y	BROWN BOVERI REVIEW, vol.71, no.5, May 1984, pages 245-250; Baden, CH H. VOGT: "Linear accelerator DYNARAY-CH20."		
A	* Whole article *	1,4 5,8	
A	US-A- 4 118 631 (FROGGATT) * Figures 1-4; column 2, line 8 - column 4, line 55 *	1,2	TECHNICAL FIELDS SEARCHED (Int. Cl.4) A 61 B A 61 N
INCOMPLETE SEARCH --- ./. The Search Division considers that the present European patent application does not comply with the provisions of the European Patent Convention to such an extent that it is not possible to carry out a meaningful search into the state of the art on the basis of some of the claims. Claims searched completely: 4-9 Claims searched incompletely: 1-3 Claims not searched: Reason for the limitation of the search: Method for treatment of the human or animal body by surgery or therapy (see art. 52(4) of the European Patent Convention).			
Place of search The Hague		Date of completion of the search 28-02-1986	Examiner CHEN
CATEGORY OF CITED DOCUMENTS X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document		T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document	



PARTIAL EUROPEAN SEARCH REPORT

EP 85 85 0224

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DOCUMENTS CONSIDERED TO BE RELEVANT			CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	
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A	<p><u>US-A- 4 045 672 (WATANABE)</u></p> <p>* Figures 1,2,5; column 1, line 65 - column 4, line 7 *</p> <p>-----</p>	4,7	<p>TECHNICAL FIELDS SEARCHED (Int. Cl.4)</p>